Validation of different accelerometers to determine mechanical loading in children

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Abstract: The purpose of this study was to assess the validity of accelerometers using force plates (i.e., ground reaction force (GRF)) during the performance of different tasks of daily physical activity in children. Thirteen children (10.1 (range 5.4–15.7) years, 3 girls) wore two accelerometers (ActiGraph GT3X+ (ACT), GENEA (GEN)) at the hip that provide raw acceleration signals at 100 Hz. Participants completed different tasks (walking, jogging, running, landings from boxes of different height, rope skipping, dancing) on a force plate. GRF was collected for one step per trial (10 trials) for ambulatory movements and for all landings (10 trials), rope skips and dance procedures. Accelerometer outputs as peak loading (g) per activity were averaged. ANOVA, correlation analyses and Bland–Altman plots were computed to determine validity of accelerometers using GRF. There was a main effect of task with increasing acceleration values in tasks with increasing locomotion speed and landing height ($P < 0.001$). Data from ACT and GEN correlated with GRF ($r = 0.90$ and 0.89, respectively) and between each other ($r = 0.98$), but both accelerometers consistently overestimated GRF. The new generation of accelerometer models that allow raw signal detection are reasonably accurate to measure impact loading of bone in children, although they systematically overestimate GRF.

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Validation of two accelerometers to determine mechanical loading of physical activities in children

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Abstract (max 200 words)

The purpose of this study was to assess the validity of accelerometers using force plates (i.e., ground reaction force (GRF)) during the performance of different tasks of daily physical activity in children.

Thirteen children (10.1 (range 5.4 – 15.7) years, 3 girls) wore two accelerometers (ActiGraph GT3X+ (ACT), GENEA (GEN)) at the hip that provide raw acceleration signals at 100 Hz. Participants completed different tasks (walking, jogging, running, landings from boxes of different height, rope skipping, dancing) on a force plate. GRF was collected for one step per trial (10 trials) for ambulatory movements and for all landings (10 trials), rope skips, and dance procedures. Accelerometer outputs as peak loading (g) per activity were averaged. ANOVA, correlation analyses and Bland-Altman plots were computed to determine validity of accelerometers using GRF.

There was a main effect of task with increasing acceleration values in tasks with increasing locomotion speed and landing height ($p<0.001$). Data from ACT and GEN correlated with GRF ($r=0.90$ and 0.89, respectively) and between each other ($r=0.98$), but both accelerometers consistently overestimated GRF.

The new generation of accelerometer models that allow raw signal detection are reasonably accurate to measure impact loading of bone in children, although they systematically overestimate GRF.

1. Introduction

Physical loading is recognized as one of the most potent modifiable lifestyle factors in the prevention of osteoporosis (Rizzoli, Bianchi, Garabedian, McKay, & Moreno, 2010). Yet, its assessment is limited to imprecise assessments using questionnaires, to laboratory conditions in terms of force plates, or to accelerometers that are validated to metabolic rather than impact force equivalents. Despite excessive validation of accelerometers against metabolic equivalents to understand the dose-response relationship between physical activity and overall or cardiovascular health (Freedson, Pober, & Janz, 2005; Pate, Almeida, McIver, Pfeiffer, & Dowda, 2006; Phillips, Parfitt, & Rowlands, 2013; Plasqui, Bonomi, & Westerterp, 2013; Reilly et al., 2006), studies using accelerometers to assess the relationship between physical loading and bone health are scarce (Garcia, Langenthal, Angulo-Barroso, & Gross, 2004; Janz, Rao, Baumann, & Schultz, 2003; Neugebauer, Hawkins, & Beckett, 2012; Rowlands & Stiles, 2012).

Recent developments of commercially available accelerometers now provide the possibility to use accelerometers as surrogate measure of physical activity not only by undefined arbitrary unit of counts that are usually processed by a smoothing integral of peak accelerations over a user defined epoch time (mostly between 1 s and 60 s) but also by determining single short signals. This newest accelerometer generation allows assessing body acceleration in raw acceleration values up to 8 g with a sampling frequency up to 100 Hz. This technical development provides new insights in bone research, where such a high resolution is needed to capture the short, high impact accelerations beneficial to bone health (Bassey & Ramsdale, 1994; Forwood, 2008; Lanyon & Rubin, 1984; Robling, Hinant, Burr, & Turner, 2002). Due to merely absent validation studies in children, the aim of this study was to validate two commercially available accelerometers against ground reaction forces using a force plate for typical tasks of physical activities of daily living in children.

2. Methods

Thirteen children (5 to16 years, three girls) were recruited to participate in the study (Table 1). All subjects were selected from a moderately active population not performing more than three hours of weekly exercise in addition to physical education at school. We excluded children involved in sports where specific plyometric training is done such as athletics, volleyball, basketball, European handball, or gymnastics. The study was approved by the local ethical committee and parents and children gave written informed consent.

All experiments took place in a laboratory (Laboratory for Movement Analysis of the Children's University Hospital of Basel) that is equipped with two force plates set flush within the floor (Kistler, Winterthur Switzerland; sampling frequency 2400 Hz). Participants were asked to restrain from strenuous exercise 24 hours prior to testing, to have slept at least seven hours, and to have had a light meal two hours before testing. Prior to the tests, children's height, mass and leg length (distance between *Spina iliaca anterior superior* and *Malleolus medialis*) were measured.

After a detailed introduction and several practice trials in order to get familiarized with the testing procedure, a series of different tasks were performed on the force plates (Table 2). The eight different tasks included walking, jogging, running, followed by landings from boxes with heights of 10, 20 and 30 cm, rope skipping, and dancing a typical breakdance move (i.e., battlerock). The ambulatory tasks were performed over a distance of 10 meters with the two force plates build in the floor at half distance. The remaining tasks were performed on one force plate. Each task was repeated 7 times (ambulatory tasks), 10 times (landings), respectively in 2-3 series of 5 (i.e., dance move) or 10 (i.e., rope skipping) times.

During testing, children wore two different commercially available triaxial accelerometers at their right hip: (1) ActiGraph GT3X+ (ACT; Actigraph, Pensacola, FL 32502, USA), (2) GeneActive (GEN; GeneActive, ActiveInsights, Kimbolton, Cambridgeshire, United Kingdom). Both accelerometers were initialized to collect triaxial data at a sampling frequency of 100 Hz. The dynamic range was ± 6 g (with a resolution of 3.9×10^{-3} g) and ± 8 g (with a resolution of 2.9×10^{-3} g) for the ACT and GEN, respectively.

ActiLife5 analysis software (version 5.10.0.0) and GENEA software (version 2.1) were used to initialize and download accelerometer data. All accelerometers were initialized to collect triaxial data at a sampling frequency of 100 Hz. To reduce the data, R project (R version 2.15.1) was used. The acceleration signals of the two accelerometers were first synchronized by the maximum cross-correlation coefficient based on their acceleration signal in the vertical direction. Then, the minimum acceleration values of the vertical axis were extracted for each trial. We excluded all values where the respective accelerometer peaked, i.e. where accelerometer values were above the maximum measurement range of 6 g (ACT) or 8 g (GEN). Maximum output values for ACT were 6.0 g, and for GEN 7.95 g. During walking, jogging and running tasks, our participants performed on average between 8-15 steps in each of the ten trials. Due to this routine, we first averaged the minimum acceleration values of each step and thereafter averaged all trials. For the landings, rope skipping and dancing tasks all minimum values were averaged. In order to test if different reduction procedures influenced the results, we also tested three other reduction algorithms: (1) reducing the data without excluding the peaked values, (2) excluding the peaked values, but only include the six trials per task with the least variance, and (3) including the six trials per task with the least variance when not excluding the peaked values. The same procedure was used to extract peak impact forces in the vertical plane (GRF) for each trial, expressed as body weights (force output (N)·(mass (kg)·acceleration due to gravity (9.81 $m\cdot s^{-2}$))$^{-1}$.

Descriptive results are given in means ± 1 standard deviation (SD) unless otherwise stated. All acceleration peaks are reported as positive vector magnitudes. A repeated measures ANOVA was used to determine differences in outputs by activity (device×activity). We calculated Pearson correlations between the GRF and the two accelerometers across all tasks for each child separately. Correlation coefficients are reported as means of the individual correlations including the 95% confidence interval based on Fisher's zr transformation. We further tested by regression analyses if sex, age, weight, height, or leg length was related to the individual correlation coefficients. A repeated measures ANOVA was used to determine differences in mean values for each activity according to the data reduction procedure (device×reduction×activity). The agreement between the methods was illustrated by Bland-Altman plots. The association between the difference and the magnitude of the measurement (i.e. heteroscedasticity) was examined by regression analysis with the difference between accelerometer and GRF as dependent and the averaged value as the independent variable. In case of heteroscedasticity, limits of agreement were recalculated using natural logarithms of accelerometer and GRF data.(Nevill & Atkinson, 1997) Subsequently, agreement was expressed as mean bias ratio multiplied and divided by the 95% agreement component (random error) on the ratio scale. All analyses were performed with Stata 11.0 and IBM SPSS Statistics 20.0. The significance level was set at 0.05.

3. Results

The characteristics of participating children are given in Table 1. Due to technical problems, one child did not perform the ambulatory tasks. Children performed on average 7.6 (standard deviation 1.3) trials for walking, jogging, and running task, respectively. Of those, 6.7 (1.5) trials had at least one step on a force plate. Children performed on average 10.6 (1.5) landings per different height, 19.5 (8.4) rope skips and 14.5 (7.2) dance moves. Means of GRF and peak vertical raw accelerometer outputs by the two accelerometers for the eight different tasks are given in Figure 1. There was a significant main effect of task ($p<0.001$) with increasing acceleration values in tasks with increasing locomotion speed and landing height. The lowest GRF values were found when walking; significantly higher values were recorded for jogging and running (1.3 (0.1) BW for walking vs. 2.2 (0.3) BW for jogging vs. 2.8 (0.4) BW for running; $p<0.001$). Landing tasks led to significantly higher GRF values than the three gait tasks ($p<0.014$). Within the landing tasks, significant higher values could be found when landing from heights of 20 and 30 cm compared to 10 cm (4.2 (0.8) BW for 10 cm vs. 5.2 (1.0) BW for 20 cm vs. 5.9 (1.2) BW for 30 cm; $p=0.001$). Rope skipping corresponded to GRF of a 10 cm landing (4.3 (0.8) BW). Independent of the task condition, GEN measured consistently higher vertical accelerations for all activities as compared to GRF, while ACT measured significantly higher values in the ambulatory tasks and the 10 cm landing, but not in the higher landings or rope skips (Table 3).

ACT did not reach its measurement limit of 6 g values for walking and jogging tasks. However, on average in 6% (range 0-21%) of all running values, 9% (0-50%) of the 10 cm-landing values, 29% (0-100%) of all 20 cm- and 30 cm-landing values, 21% (0-65%) of rope skipping values, and 2% (0-13%) of all dancing values had to be omitted due to peaking. GEN outputs reached the measurement limit of 8 g in 14% (0-29%) of all running values, in 28% (0-91%) of all 10 cm-landings, in 48% (0-100%) of all 20 cm-landings, in 62% (0-100%) of all 30cm-landings, in 22% (0-91%) of all rope skipping values, and in 9% (0-27%) of dancing values. Despite these exclusions, each child had at least 4 valid peak measures for the ambulatory activities. After omission of these peaked values the correlation coefficient between GRF and ACT over all tasks was r=0.90 (95% confidence interval: 0.68 – 0.97), between GRF and GEN r=0.89 (0.66 – 0.97), and between the two accelerometers r=0.98 (0.92 – 0.99). Sex, age, weight, height, and leg length of the children did not have a significant influence on the correlation coefficients. These correlation coefficients were similar for all data reduction procedures (data not shown). There were no major differences of mean values by device and activity according to different data reduction procedures used, i.e. whether peaked values were included or not or whether all peaks per activity or only the values with the least variance were used for analyses (Table 3). Single statistically significant differences of means occurred for GEN and ACT outputs of the higher impact activities when peaked values were included versus excluded. Figure 2 shows the Bland-Altman plots for GRF versus ACT and GRF versus GEN, respectively. Systematic biases (i.e. mean difference) between GRF-ACT and GRF-GEN were 0.46 g and 1.39 g, respectively, with limits of agreement (±2 SD) ranging from -1.15 g to 2.07 g and -0.45 g to 3.24 g, respectively. Moreover, the differences of means increased with higher speeds and impacts. In both cases, there was a significant association between the difference and the magnitude of the measurement ($p<0.001$). To reduce heteroscedasticity errors, transformation of accelerometer and GRF data into natural logarithms gave a mean bias ratio of 1.18 ×/÷1.57 for ACT versus GRF and 1.43 ×/÷1.59 for GEN versus GRF, respectively.

4. Discussion

As accelerometers are perfectly suited to measure impact loading of bone, their calibration against GRF is a prerequisite for their use in field studies focusing on mechanical loading of bone. This study provides the first validation study in children of two accelerometers capable of measuring raw accelerations with sufficient precision against GRF. The main finding was that, despite the high correlation coefficients between the applied methods, both accelerometers systematically and significantly overestimated directly measured GRF. Moreover, measurement bias increased with higher loadings.

Availability of raw acceleration data from commercially available accelerometers such as ACT and GEN have the potential to measure and quantify impact loading of bone over days in observational and intervention studies. Results from this study indicate that both accelerometers are reasonably accurate in determining GRF applied to the skeleton and that correlation of these two accelerometers is high.

Although already accelerometer data of former device generations showed good correlations to force platform outputs (Servais, Webster, & Montoye, 1984), validation studies to assess the relationship between physical loading and bone health are still scarce. Except for one study in adults (Rowlands & Stiles, 2012), previous accelerometer-based validation studies did not use appropriate devices that were able to capture single raw acceleration signals as output with an sufficiently high sampling rate for the adequate measurement of very short impact accelerations. These studies (Garcia, et al., 2004; Janz, et al., 2003; Neugebauer, et al., 2012) used accelerometers with epoch times of 15 sec and sampling rates of 40 Hz which questions results for several reasons. First, the rate of data acquisition is determined by the sampling frequency of the monitor. To ensure that all human movements are adequately measured, the sampling frequency must be at least twice the speed of the fastest movement (Chen & Bassett, 2005). Most commercially available accelerometers are sampling between 1 to 64 Hz, including the ones used in the validation studies mentioned above (Garcia, et al., 2004; Janz, et al., 2003; Neugebauer, et al., 2012). They are appropriate to assess frequencies for normal non-impact physical activities in humans that are generally below 8 Hz (Winter, Quanbury, & Reimer, 1976). However, their use is not meaningful for determining impact-related accelerations. The signal is cut off before it reaches its true maximum and this results in inappropriately low measured values. Second, accelerometer outputs are usually filtered by a band pass filter which attenuates all frequencies outside a set range (0.25 – 7 Hz) to increase linearity of the output and decrease artifacts. A restricted bandwidth might lead to an underestimation of light physical activities or the leveling off for vigorous activities including high impact loading (Chen & Bassett, 2005). Third, the raw output of traditional accelerometers to measure physical activity is given in so-called "counts" that is an arbitrary unit of unknown physical or physiological meaning. The original voltage signal of the accelerometer, after being filtered and amplified, is then sampled at a prefixed frequency and converted to the digital "raw counts." These raw counts are usually converted to the final counts by a method that uses the area under the curve algorithm by integrating or averaging the signals over a predefined time window of 1 sec to 1 min, so called epoch time. Especially when short bursts of high intense physical activities occur with low intense activities during the same epoch time, the data will be averaged and presented as intermediate intensity (Chen & Bassett, 2005). And fourth, the former generation of accelerometers had a dynamic range of ± 2 g. Thus, all activities with accelerations above 2 g could not be accurately assessed. All in all, our accelerometers were not affected by these technical restrictions. They provided raw accelerometer signals at a sampling frequency of 100 Hz and they were able to capture a dynamic range of ± 6-8 g.

It has frequently been shown that programmes incorporating regular weight-bearing exercise can result in 1% to 8% improvements in bone strength at the loaded skeletal sites in children and adolescents (Hind & Burrows, 2007; Nikander et al., 2010). Nevertheless, there is also a number of contradictive studies available that questioned the large training-induced increases in bone strength. These latter studies, however, were often based on relatively small sample sizes, short follow-up periods, the measurements of different skeletal sites using different imaging techniques, and an imprecise assessment of types and doses of training (Nikander, et al., 2010). Our present study presents a method to precisely quantify mechanical loading of bone by a valid objective assessment over time. Previously, this assessment was limited to questionnaire- and interview-based data that determined intensity of loading by different activities (Ainsworth, Shaw, & Hueglin, 2002; Kemper, Bakker, Twisk, & van Mechelen, 2002) with similar peak strains as we found (Groothausen, Siemer, Kemper, Twisk, & Welten, 1997). Yet, questionnaires and interviews have limited reliability and validity especially when used in children (Sallis, 1991). With our objective assessment, we may be able to clear the contradictory results of earlier studies and to determine dose-response patterns between mechanical loading and bone health in children as already elegantly done in adults (Vainionpaa et al., 2007).

While this study provides insight in the use of accelerometers to estimate GRF in children, several limitations must be addressed. Our accelerometers captured peak accelerations up to 6 g (ACT) and 8 g (GEN), which limits their use for very high impact loadings (e.g., plyometrics). This upper limit of measurement range may not be relevant for the adaptation of bone to loading. Studies in adults demonstrated that GRF as little as 2 g (Vainionpaa, et al., 2007), but certainly those forces above 4 g (Vainionpaa et al., 2006) were sufficient to induce beneficial structural changes of bone strength. Only vertical forces and accelerations were assessed, which may not be adequate for certain more variable physical activities of daily living for which resultant GRF may be more appropriate. Moreover, artifact detection relevant in free-living conditions was not studied in the current study as it was not necessary in our highly controlled laboratory setting. Both accelerometers systematically overestimated GRF, irrespective of individual characteristics of the children. Based on the lower limits of agreement among accelerometer measures and GRF, one has to consider that the minimal osteogenic value of 4 g (Vainionpaa, et al., 2006) is in the worst case scenario attained with ACT and GEN outputs of 3.01 g and 3.61 g, respectively. To correct for this bias, a device specific regression model may be established to report GRF based on accelerometer outputs as done previously (Neugebauer, et al., 2012). For this purpose, a larger study population with equal gender distribution is needed.

5. Conclusion

The new generations of accelerometer models that allow raw signal detection at a high sampling rate are reasonably accurate to measure impact loading of bone in children, although they systematically overestimate GRF. Future devices should increase the measurement range for higher accelerations. Furthermore, regression models to determine peak GRF based on raw acceleration outputs for youth may be developed.

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TABLE 1: Characteristics of the participants (3 girls, 10 boys).

Variable	Mean (Standard deviation)
Number of children (n)	13
Age (yr)	10.1 (3.0)
Mass (kg)	36.2 (15.0)
Height (cm)	143.7 (18.8)
Body Mass Index (kg/m^2)	16.8 (2.6)
Leg length (cm)	74.0 (11.5)

TABLE 2: Description of the different tasks

Task	Description	Number of performed trials
Walking	Walking at 3 km/h	7 trials
Jogging	Jogging at 6 km/h	7 trials
Running	Running at 10 km/h	7 trials
Landings 10cm	Landings from a 10 cm box	2 series with 5 jumps
Landings 20cm	Landings from a 20 cm box	2 series with 5 jumps
Landings 30cm	Landings from a 30 cm box	2 series with 5 jumps
Dance Move	Breakdance move	2-3 series with 5 moves
Rope Skipping	Free rope skipping	2-3 series with 10 skips

TABLE 3: Peak impact forces by force plate and peak accelerations by accelerometers according to different activities and different reduction procedures. Values are means ± standard errors.

	Peaked values excluded				Peaked values included			
	Mean over all trials		Mean over trials with the least variance		Mean over all trials		Mean over trials with the least variance	
	N included trials/child	**Mean (SD)**	**N included trials/child**	**Mean (SD)**	**N included trials/child**	**Mean (SD)**	**N included trials/child**	**Mean (SD)**
Walking (n=12 children)								
GRF (BW)	7.2 (1.3)	1.33 (0.13)	6.0 (0)	1.35 (0.09)	7.2 (1.3)	1.33 (0.13)	6.0 (0)	1.35 (0.09)
Actigraph (g)	7.7 (1.1)	1.70 (0.13)‡	6.0 (0)	1.65 (0.13)‡	7.7 (1.1)	1.70 (0.13)‡	6.0 (0)	1.65 (0.12)‡
GeneActive (g)	7.7 (1.1)	1.73 (0.19)‡	6.0 (0)	1.69 (0.18)‡	7.7 (1.1)	1.73 (0.19)‡	6.0 (0)	1.69 (0.18)‡
Jogging (n=12 children)								
GRF (BW)	7.0 (1.8)	2.20 (0.31)	5.9 (0.3)	2.32 (0.24)	7.0 (1.8)	2.20 (0.31)	5.9 (0.3)	2.32 (0.24)
Actigraph (g)	8.3 (1.5)	2.69 (0.31)‡	6.0 (0)	2.51 (0.27)	8.3 (1.5)	2.69 (0.32)‡	6.0 (0)	2.51 (0.27)
GeneActive (g)	8.3 (1.5)	3.06 (0.51)‡	6.0 (0)	2.80 (0.50)‡	8.3 (1.5)	3.08 (0.53)‡	6.0 (0)	2.80 (0.50)‡
Running (n=12 children)								
GRF (BW)	5.8 (1.5)	2.76 (0.42)	5.4 (0.8)	2.32 (0.40)	5.8 (1.5)	2.76 (0.42)	5.4 (0.8)	2.32 (0.40)
Actigraph (g)	7.0 (1.3)	3.98 (0.36)‡	6.0 (0)	3.82 (0.56)‡	7.0 (1.3)	4.08 (0.44)‡	5.9 (0.3)	4.07 (0.78)‡
GeneActive (g)	7.0 (1.3)	4.76 (0.62)‡	6.0 (0)	4.42 (0.74)†‡	7.0 (1.3)	5.17 (0.80)†‡	5.9 (0.3)	5.02 (1.36)‡
Landings 10cm (n=13 children)								
GRF (BW)	10.6 (1.4)	4.21 (0.79)	6.0 (0)	4.03 (0.98)	10.6 (1.4)	4.21 (0.79)	6.0 (0)	4.03 (0.98)
Actigraph (g)	9.7 (2.3)	4.68 (0.49)‡	6.0 (0)	4.70 (0.60)‡	10.7 (1.7)	4.75 (0.56)‡	6.0 (0)	4.83 (0.76)‡
GeneActive (g)	7.7 (3.4)	5.99 (0.87)‡	5.3 (1.5)	5.99 (1.06)‡	10.7 (1.7)	6.38 (1.05)‡	6.0 (0)	6.55 (1.46)‡
Landings 20cm (n=13 children)								
GRF (BW)	10.4 (0.8)	5.22 (1.05)	6.0 (0)	5.26 (1.08)	10.4 (0.8)	5.22 (1.05)	6.0 (0)	5.26 (1.08)

Actigraph (g)[a]	7.3 (3.6)	5.09 (0.57)	5.1 (1.9)	5.19 (0.51)	10.4 (0.8)	5.24 (0.64)	6.0 (0)	5.37 (0.64)†
GeneActive (g)[b]	5.3 (3.7)	6.54 (0.87)‡	4.1 (2.5)	6.59 (0.94)‡	10.4 (0.8)	7.02 (0.93)‡	6.0 (0)	7.24 (1.00)‡
Landings 30cm (n=13 children)								
GRF (BW)	10.8 (1.5)	5.94 (1.18)	6.0 (0)	5.79 (1.04)	10.8 (1.5)	5.94 (1.18)	6.0 (0)	5.79 (1.04)
Actigraph (g)[a]	7.9 (3.7)	5.35 (0.39)	5.4 (1.7)	5.35 (0.48)	10.9 (1.9)	5.5 (0.42)	6.0 (0)	5.62 (0.56)
GeneActive (g)[b]	4.2 (3.8)	6.25 (0.56)‡	3.5 (2.9)	6.30 (0.59)‡	10.9 (1.9)	7.24 (0.84)†‡	6.0 (0)	7.52 (0.90)†‡
Rope skipping (n=13 children)								
GRF (BW)	16.2 (5.7)	4.29 (0.75)	6.0 (0)	4.41 (0.70)	16.2 (5.7)	4.29 (0.75)	6.0 (0)	4.41 (0.70)
Actigraph (g)	15.2 (8.0)	4.64 (0.56)	5.9 (0.3)	4.81 (0.72)	19.5 (8.4)	4.86 (0.69)	6.0 (0)	5.07 (0.88)†‡
GeneActive (g)	15.7 (9.7)	5.75 (0.92)‡	5.6 (1.4)	5.87 (1.02)‡	19.5 (8.4)	6.02 (1.12)‡	6.0 (0)	6.38 (1.38)‡
Dance move (n=13 children)								
GRF (BW)	7.2 (2.8)	2.43 (0.87)	5.4 (1.5)	2.43 (0.87)	7.2 (2.8)	2.43 (0.87)	5.4 (1.5)	2.43 (0.87)
Actigraph (g)	14.2 (7.0)	3.59 (1.16)‡	5.9 (0.3)	3.59 (1.16)‡	14.2 (6.9)	3.59 (1.17)‡	5.9 (0.3)	3.59 (1.17)‡
GeneActive (g)	13.0 (6.2)	4.37 (1.54)‡	5.9 (0.3)	4.37 (1.54)‡	14.2 (6.9)	4.58 (1.60)‡	5.9 (0.3)	4.58 (1.60)‡

GRF: ground reaction forces in body weights (BW). Mean over trials with the least variance in each activity are the six trials with the least variance. † refers to significant difference in data reduction compared to mean over all trials with excluded peaked values; ‡ refers to significant difference in device compared to GRF. [a] when peaked values were excluded, Actigraph data for Landing 20cm and Landing 30cm are based on n=12 children. [b] when peaked values were excluded, GeneActive data for Landing 20cm are based on n=11 children, and Landing 30cm on n=8 children, respectively.

Figure captions

Figure 1: Ground reaction forces and vertical peak accelerations by accelerometers according to different activities. Values are means ± standard error.

Figure 2: Comparison of ground reaction forces (GRF) by force plate and raw vertical peak accelerations by (A) Actigraph GT3X+ and (B) GeneActive by the Bland-Altmann plots. Walk/Jog/Run (☐), Dance Move/Rope Skipping (▲), Landings (◆). Solid lines represent the mean difference between the accelerometer and ground reaction force. Upper and lower dashed lines represent the 95% limits of agreement (mean difference ± 2 SD of the difference).

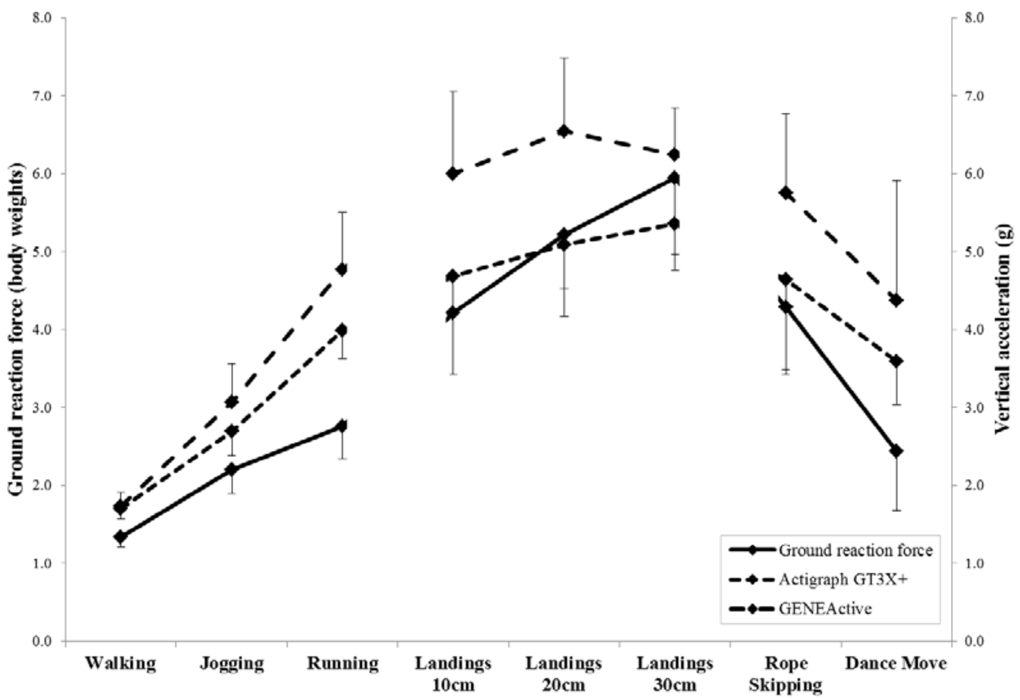
Ground reaction force (body weights)
Vertical acceleration (g)
0.0
1.0
2.0
3.0
4.0
5.0
6.0
7.0
8.0
Walking
Jogging
Running
Landings 10cm
Landings 20cm
Landings 30cm
Rope Skipping
Dance Move
Ground reaction force
Actigraph GT3X+
GENEActive

Figure 1

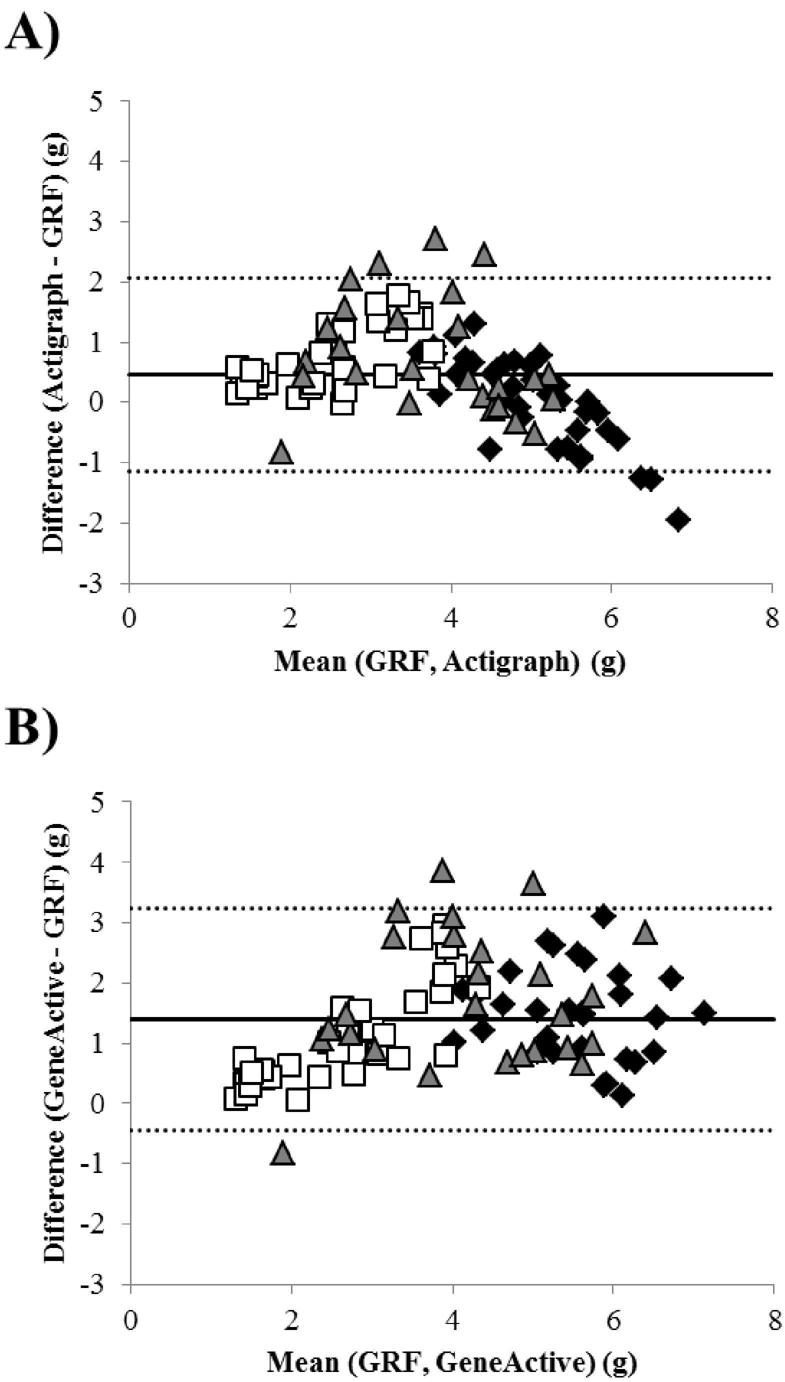
A)
Difference (Actigraph - GRF) (g)
5
4
3
2
1
0
-1
-2
-3
0
2
4
6
8
Mean (GRF, Actigraph) (g)
B)
Difference (GeneActive - GRF) (g)
5
4
3
2
1
0
-1
-2
-3
0
2
4
6
8
Mean (GRF, GeneActive) (g)

Figure 2